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SPECIFICATION

MEASURING INSTRUMENT PROVIDED WITH SOLID COMPONENT

CONCENTRATING MEANS

TECHNICAL FIELD

The present invention relates to a measuring instrument to be used for measuring the concentration of a specific component (such as glucose or cholesterol) in a sample liquid such as blood.

BACKGROUND ART

Some methods of measuring the concentration of a specific component in a sample liquid employ electrochemical techniques for example. In these methods, a reaction system is constructed from a sample liquid, an oxidoreductase and an electron transporter for example, voltage is applied to this reaction system using an electrode, and the concentration of the specific component is computed based on the resulting response current. Such a reaction system is constructed for example on a biosensor provided with a reagent part comprising an oxidoreductase and an electron transporter. Because an oxidation-reduction reaction occurs between the specific component and the electron transporter in the reaction system due to the catalytic effect of the oxidoreductase, the amount of the electron transporter

electron movement between the reduced (or oxidized) electron transporter occurring in the reaction system and an electrode. Consequently, the accuracy of concentration measurement is greatly affected by the accuracy of measurement of response current.

In such methods, when blood (blood containing blood cells) is used as the sample liquid electron movement between the electrode and the electron transporter is impeded by blood cells present on the surface of the electrode. As a result, the measured response current value is reduced as the number of blood cells increases, producing measurement errors. Moreover, if the proportion of blood cells in the blood (hematocrit value) is different the measured response current will be different even if the glucose concentration is the same

To solve these problems, methods have been proposed of separating blood cells from blood in the measuring instrument. Methods of separating blood cells include for example a method of providing a separation membrane in that part of the measuring instrument where the blood or other sample liquid is introduced (see for example Japanese Patent Application Laid-open No. H8-114539 and Patent Publication No. 2002-508698), and a method of covering the surface of the electrode with a polymer membrane (see for example Japanese Patent Application Laid-open No. H6-130023, Japanese Patent Application Laid-open No. H9-243591 and Japanese Patent Application Laid-open No. 2000-338076).

However, because in methods which filter blood cells in the measuring instrument the plasma component must pass through the separation membrane it takes a long time for the plasma to reach the surface of the electrode, prolonging the measurement time. This problem can be solved by securing a large quantity of whole blood to be used, but this places a heavy burden of blood collection on the user.

DISCLOSURE OF THE INVENTION

It is an object of the present invention to control the effect of the solid component in the sample liquid and precisely measure concentration with a small amount of sample liquid while keeping the measurement time short.

In the present invention a measuring instrument is provided which is an analytical instrument comprising a channel for moving a sample liquid containing a solid component and for providing a liquid reaction field and a first and second electrode which are used to apply voltage to the liquid reaction field, wherein the first electrode has an electron transfer interface for providing electrons to the liquid reaction field or receiving electrons from the reaction field when voltage is applied to the liquid reaction field via the first and second electrodes, and which is provided with concentration means for increasing the concentration of solid components in that part which contacts the electron transfer interface in the liquid reaction field.

The concentration means comprises for example a water-absorbing layer containing an absorbent polymer material. The absorbent polymer material must be one capable of absorbing enough liquid component to achieve the object of the present invention, and the amount of water absorbed must not be such as to affect the measurement results. Consequently, it is desirable to use an absorbent polymer material having a water absorption power of 10 to 500 g/g.

The measuring instrument is provided for example with a substrate on which are formed the first and second electrodes and with a cover laid over the substrate.

The water-absorbing layer is formed as a film on at least that part of the cover which faces the electron transfer interface for example. In this case, the water-absorbing layer is preferably formed so that its dimension in the direction of thickness of the substrate without water absorption and with water absorption is $1/30$ to $1/10$ and $1/5$ to $3/5$, respectively, of the dimension in the direction of thickness of the channel. The water-absorbing layer can also be made water-soluble.

The water-absorbing layer may also be formed across all or most of the length of the channel. Such a water-absorbing layer can be the cover if the cover is formed so as to include an absorbent material.

The water-absorbing layer can also be constructed by supporting a powder comprising an absorbent polymer material on the cover. The weight average grain size of the powder without water absorption is 100 to 1000 μm for

example. This is because if the average grain size is unsuitably small it is difficult to form the water-absorbing layer so that it can absorb enough water to achieve the object, while if the average grain size is unsuitably large it will impede the flow of water in the channel more than necessary.

The water-absorbing layer can also be placed downstream in the flow of sample liquid from the electron transfer interface in the channel. This water-absorbing layer is placed for example on either the substrate or the cover. In this case the dimension of the water-absorbing layer in the direction of flow of the sample liquid is preferably $1/4$ to $1/2$ of the distance from the channel inlet to the furthest downstream point of the electron transfer interface in the direction of flow of the sample liquid so that the solid component can be concentrated as intended. For the same reasons it is desirable that during water absorption the thickness of the part having the formed water-absorbing layer be 0 to 15 μm .

The water-absorbing layer can also be formed so as to have a part formed in at least one of a location upstream from and adjacent to the electron transfer interface and a location downstream from and adjacent to the electron transfer interface. In this case, the water-absorbing layer is preferably formed so as to have both a part formed in a location upstream from and adjacent to the electron transfer interface and a part downstream from and adjacent to the electron transfer

interface, and it is formed for example so as to surround the electron transfer interface.

The concentrating means can also be provided downstream in the channel from the electron transfer interface in the direction of movement of the sample liquid and can be formed by a absorption-resistant dam to impede the movement of the solid components.

The dam is formed so that the thickness of the channel where the dam part is formed is 5 to 15 μ m for example so as to concentrate the solid components as intended.

The same liquid which is measured in the measuring instrument of the present instrument may classically be blood containing blood cells as the solid component. The measuring instrument of the present invention can be used for a wide range of sample liquids containing solid components, and the sample liquid which is measured is not limited to blood.

BRIEF DESCRIPTION OF THE DRAWINGS

Figure 1 is a total oblique view of a biosensor according to the first embodiment of the present invention.

Figure 2 is an exploded oblique view of the biosensor shown in Figure 1.

Figure 3 is a cross-section along line III-III in Figure 1.

Figures 4A and 4B are cross-sections as in Figure 3 for explaining the flow of blood through the interior channel of the biosensor.

Figure 5 is a typical view showing the biosensor shown in Figure 1 mounted on a concentration measuring system.

Figure 6 is a cross-section showing a biosensor according to the second embodiment of the present invention as in Figure 3.

Figure 7 is a cross-section showing a biosensor according to the third embodiment of the present invention as in Figure 3.

Figures 8A and 8B are cross-sections showing a biosensor according to the fourth embodiment of the present invention as in Figure 3.

Figure 9 is a cross-section showing a biosensor according to the fifth embodiment of the present invention as in Figure 3.

Figures 10A and 10B are oblique views showing the biosensor of Figure 9 with the cover and spacers removed.

Figure 11 is a graph of the time course of response current in the biosensor of Example 1.

Figure 12 is a graph of the time course of response current in the biosensor of Comparative Example 1.

Figure 13 is a graph showing the effect of hematocrit (Hct) in the biosensor of Example 1.

Figure 14 is a graph showing the effect of hematocrit (Hct) in the biosensor of Comparative Example 1.

BEST MODE FOR CARRYING OUT THE INVENTION

Embodiments 1 through 5 of the present invention are explained in detail below with reference to the drawings.

First, the first embodiment of the present invention is explained using blood sugar measurement as an example with reference to Figures 1 through 5.

The biosensor 1 shown in Figures 1 and 2 is used to measure glucose concentrations in blood, and is used mounted on concentration measuring system 2 (see Figure 5). This biosensor 1 has a configuration in which cover 5 is laid over rectangular substrate 3 with a pair of sensors 40 and 41 in between, and as shown in Figure 3, capillary 6 is formed by these elements 3, 40, 41 and 5.

Capillary 6 moves blood by means of capillary action, and has internal channel 60 for holding the blood. This internal channel 60 extends in the short direction of substrate 3, and communicates with the outside via end openings 61 and 62. End opening 61 is used to introduce blood into internal channel 60, and when blood is moved through internal channel 60 end openings 62 is used to exhaust gas from internal channel 60.

As shown in Figures 1 through 3, the pair of spacers 40 and 41 serve to join cover 5 to substrate 3 and to determine the dimensions of internal channel 60 of capillary 6. The pair of spacers 40 and 41 extend in the short direction of substrate 3, and are arranged with a gap between them in the long direction of substrate 3.

Working electrode 31 and counter electrode 32 are formed extending in the long direction of substrate 3 on upper surface 30 of substrate 3. Reagent site 33 is also provided on upper surface 30 of substrate 3 so as to successively transect both working electrode 31 and counter electrode 32. The part of working electrode 31 which contacts reagent site 33 comprises electron transfer interface 31a.

Reagent site 33 is formed as a solid which comprises an oxidoreductase and an electron transporter. Glucose oxidase or glucose dehydrogenase for example can be used as the oxidoreductase. The electron transporter is one that is oxidized or reduced by application of voltage and reactions, and in blood sugar measurement potassium ferricyanide for example is used as the electron transporter. In this embodiment, the electron transporter is included in its oxidized form before the blood is supplied.

As shown in Figures 2 and 3, water-absorbing layer 51 is formed on one side 50 of cover 5. This water-absorbing layer 51 is formed on one side 50 of cover 5 so as to face electron transfer interface 31a, which is located in internal channel 60 on working electrode 31. This water-absorbing layer 51 can be formed by affixing a water-absorbing sheet comprising an absorbent polymer material to cover 5. This water-absorbing layer 51 is formed so that its thickness dimension is for example $1/30$ to $1/10$ of the height (H) of internal channel 60 without water absorption, and so that its thickness

dimension with water absorption is $1/5$ to $3/5$ the height (H) of internal channel 60.

A material with water absorption power of 10 to 500 g/g for example is used as the absorbent polymer material. Specific examples of the absorbent polymer material include crosslinked acrylate polymers, crosslinked vinyl alcohol-acrylate copolymers, crosslinked maleic anhydride grafted polyvinyl alcohol, crosslinked isobutylene-maleic anhydride copolymer, carboxymethylcellulose crosslinked with alkali salts, crosslinked partially neutralized polyacrylic acid and the like. All of water absorbing layer 51 may be formed of an absorbent polymer material, or it may be formed as a layer which is a mixture of an absorbent polymer material and another, non-absorbent polymer material. Water-absorbing layer 51 can be formed by first applying a solution of an absorbent polymer material dissolved in a solvent to cover 5 and then drying it.

As shown in Figure 5, concentration measuring system 2 is equipped with first and second terminals 20a and 20b, voltage applying part 21, current measuring part 22, detection part 23, control part 24, computation part 25 and display part 26.

First and second terminals 20a and 20b are for purposes of contact with terminals 31b and 32b of working electrode 31 and counter electrode 32 in biosensor 1 when biosensor 1 is mounted on concentration measuring system 2.

Voltage applying part 21 is for applying voltage between working electrode 31 and counter electrode 32 of biosensor 1 via first and second terminals 20a and 20b. Voltage applying part 21 is electrically connected to first and second terminals 20a and 20b. Voltage applying part 21 is configured for example with a direct current power source such as a dry battery or charger.

Current measuring part 22 is for measuring current values when voltage is applied by means of voltage applying part 21 between terminals 31b and 32b of working electrode 31 and counter electrode 32.

Detection part 23 is for detecting based on the current values measured by current measuring part 22 whether or not sample liquid has been supplied to reagent part 33 (see Figures 1 through 3) after biosensor 1 has been mounted on concentration measuring system 2.

Control part 24 is for controlling voltage applying part 21 and selecting whether voltage is applied or is not applied between working electrode 31 and counter electrode 32.

Computation part 25 is for computing blood sugar values according to current values measured by current measuring part 22. Computation part 25 is constructed so as to compute blood sugar values by amperometric means for example.

Detection part 23, control part 24 and computation part 25 can be constructed for example by connecting multiple memories (such as ROM or RAM) to one CPU.

Display part 26 is for displaying the computation results of computation part 25 along with error messages and operating procedures for example, and is formed for example from a liquid crystal display device.

Next, the procedures for measuring blood sugar using biosensor 1 and concentration measuring system 2 are explained.

As clearly shown in Figure 5, biosensor 1 is first set on concentration measuring system 2. Terminals 31b and 32b of working electrode 31 and counter electrode 32 of biosensor 1 are thus connected to first and second terminals 20a and 20b of concentration measuring system 2. This allows voltage to be applied between working electrode 31 and counter electrode 32 via first and second terminals 20a and 20b. Under actual measurement conditions, constant voltage is applied between working electrode 31 and counter electrode 32 as soon as biosensor 1 is mounted on concentration measuring system 2. The constant voltage applied between working electrode 31 and counter electrode 32 is set for example in the range of 100 to 1000 mV. In this embodiment, application of constant voltage between working electrode 31 and counter electrode 32 is performed continuously until the response current for computing blood sugar has been measured.

Next, blood is supplied via end opening 61 of biosensor 1. As shown in Figures 4A and 4B, blood BL travels by capillary action through internal channel 60 from end opening 61 to end opening 62 of capillary 6. As

clearly shown in Figure 4B, blood BL is introduced until blood BL reaches end opening 62 and internal channel 60 of capillary 6 is full of blood BL. In this process reagent site 33 (see Figure 4A) is dissolved by blood BL and a liquid reaction system is constructed in internal channel 60. At this time in water-absorbing layer 51 the plasma component of blood BL is absorbed by water-absorbing layer 51, which grows in thickness. In this way, the movement of blood cells B1 is impeded by water-absorbing layer 51, and the concentration of blood cells B1 rises on and around the surface of electron transfer interface 31a of working electrode 31.

Within the liquid reaction system, the glucose in blood BL is oxidized by oxidoreductase while the electron transporter becomes reduced. When voltage is applied, the reduced electron transporter moves to the surface of electron transfer interface 31a of working electrode 31, supplies electrons to electron transfer interface 31a and reverts to being an oxidized electron transporter. The amount of electrons supplied to electron transfer interface 31a is measured as response current by current measuring part 22 via first and second terminals 20a and 20b.

Meanwhile, the response current measured by current measuring part 22 is monitored by detection part 23, and once the response current crosses a threshold detection part 23 detects that blood has been supplied to reagent part 33 and reagent part 33 has dissolved. When detection part 23 has detected that blood has been

supplied, detection part 23 then evaluates whether a fixed time has passed since this detection.

When detection part 23 has determined that a fixed time has passed, response current is measured at current measuring part 22 and blood sugar is computed by computing part 25 based on these response current values. Computation of blood sugar is performed by first converting response current values to voltage values and then comparing these voltage values to a previously-prepared working curve showing the relationship between voltage values and blood sugar values. The computation results of computation part 25 are displayed on display part 26 for example.

In this embodiment, when blood BL is supplied to internal channel 60 of capillary 6 the plasma component of blood BL is absorbed by water-absorbing layer 51, increasing the concentration of blood cells B1 on and around the surface of electron transfer interface 31a of working electrode 31. In this way, the area on and around the surface of electron transfer interface 31a is artificially in the same state as if high-hematocrit blood BL were being supplied. Moreover, if an absorbent polymer material with a water absorption power of 10 to 500 g/g is used, water-absorbing layer 51 will absorb more plasma the lower the hematocrit value of blood BL. As a result, a similar high hematocrit state can be achieved around water-absorbing layer 51 regardless of whether the hematocrit value is high or low.

Biosensor 1 can also resolve issues involved in separating blood cells from blood in the measuring instrument. That is, in methods of separating blood cells the plasma component has to be passed through a separation membrane, prolonging the measurement time and reducing the amount of blood that can be used in the reaction relative to the amount supplied. In contrast, because in biosensor 1 there is no separation membrane or other obstruction to the progress of blood BL through internal channel 60 of capillary 6, measurement time is not prolonged as it is with a separation membrane. Moreover, because in biosensor 1 most of the blood BL supplied to internal channel 60 of capillary 6 can be reacted with oxidoreductase of reagent part 33, concentrations can be measured properly using only a tiny amount of blood BL.

Next, a biosensor according to the second embodiment of the present invention is explained with reference to Figure 6.

In the biosensor 1A shown in Figure 6, water-absorbing layer 51A is formed across the entire length of capillary 6. This water-absorbing layer 51A can be formed for example by affixing a water-absorbing sheet formed from an absorbent polymer material to the cover. Water-absorbing layer 51A can also be formed by coating the cover with a solution of an absorbent polymer material dissolved together with an adhesive component in a solvent, and then drying the coat.

An absorbent layer can also be formed across the entire length of capillary 6 by making all of cover 5 water absorbent and having all of cover 5 comprise the water-absorbing layer. Such a water-absorbing layer (cover) can be formed for example by first mixing an absorbent polymer material with a resin material to make a molding material which is then subjected to resin molding.

Next, a biosensor according to the third embodiment of the present invention is explained with reference to Figure 7.

The biosensor 1B shown in Figure 7 is constructed with water-absorbing layer 51B having absorbent polymer grains. This water-absorbing layer 51B has a construction in which absorbent polymer grains 51Bb are held on one side of double-sided tape 51Ba, and is affixed to the cover using the adhesiveness of the other side of the double-sided tape. An absorbent polymer with a weight-average grain size of 100 to 1000 μm is used by preference.

Next, a biosensor according to the fourth embodiment of the present invention is explained with reference to Figures 8A and 8B.

In the biosensor 1C shown in Figure 8A, water-absorbing layer 51C is formed on substrate 3 downstream in the direction of blood flow from electron transfer interface 31a of working electrode 31. However, water-absorbing layer 51C may also be formed on cover 5.

In this biosensor 1C, as shown in Figure 8B, when blood BL is introduced into capillary 6 water-absorbing layer 51C expands, decreasing the spatial cross-sectional area of the part of capillary 6 having formed water-absorbing layer 51C. As a result, the movement of blood cells B1 is impeded by water-absorbing layer 51C, blood cells B1 accumulate near electron transfer interface 31a and the concentration of blood cells B1 increases around electron transfer interface 31a.

In order for this to be effective, water-absorbing layer 51C is preferably formed so that the distance L between water-absorbing layer 51C and the upper surface of the capillary is 0 to 15 μm when capillary 6 is filled with blood. Moreover, in order to more reliably increase the concentration of blood cells B1 around electron transfer interface 31a it is desirable that dimension W1 of water-absorbing layer 51C in the direction of flow of blood BL be made relatively large. Dimension W1 in this case is preferably set to about 1/4 to 1/2 of the distance W2 between inlet 68 of capillary 6 and downstream end 31a' of electron transfer interface 31a.

A function similar to that of the water-absorbing layer 51C shown in Figures 8A and 8B can be achieved with a non-(or low-) water-soluble dam part. That is, rather than causing blood cells to accumulate due to absorption of the plasma component and expansion, the cross-sectional dimensions downstream from electron transfer interface 31a in capillary 6 can be made small by the formation of a dam before blood is supplied. This dam is

preferably formed so that the distance (corresponding to L in Figure 8B) between the dam and the substrate (or cover) is 5 to 15 μm . The dam can be formed on either the substrate or the cover.

Next, a biosensor according to the fifth embodiment of the present invention is explained with reference to Figures 9, 10A and 10B.

In the biosensor 1D shown in Figure 9, water-absorbing layer 51D is formed so as to have a part adjoining electron transfer interface 31a of working electrode 31. As shown in Figure 10A, water-absorbing layer 51D may be arranged in two sites both upstream and downstream from electron transfer interface 31a (see Figure 9), or as shown in Figure 10B it can be formed as a rectangular frame. In the configuration shown in Figure 10A, one of the two water-absorbing layers 51D may also be omitted.

The measurement of blood glucose concentrations was taken as an example in the above explanation, but the present invention is applicable to measurement of other components in blood such as cholesterol, lactic acid, bilirubin and the like, as well as to sample liquids other than blood.

Examples

It is shown below that the effect of blood cells in blood on measurement of response current is reduced by the biosensor of the present invention.

[Example 1]

(Preparation of glucose sensor)

In this example, a biosensor was formed with the same configuration as in Figures 1 through 4. In this biosensor, the length L, width W and height H of internal channel 60 of capillary 6 are given as 3 mm, 1 mm and 40 μm , respectively (see Figures 1 and 3). Working electrode 31 and counter electrode 32 were formed by screen printing using carbon ink (Japan Acheson "Electrodag 423SS"). Reagent site 33 was given a two-layer structure consisting of an electron transport layer and an enzyme-containing layer. The electron transport layer was formed by first applying 0.4 μL of a first material liquid comprising an electron transporter to substrate 3, and then blow drying (30°C, 10% Rh) the first material liquid. The enzyme-containing layer was formed by first applying 0.3 μL of a second material liquid containing oxidoreductase to the electron transport layer, and then blow drying (30°C, 10% Rh) the second material liquid.

The first material liquid was prepared by mixing the liquid components in the following table 1 in numerical order from 1 to 4, letting the mixture sit for 1 to 3 days and then adding an electron transporter thereto. The second material liquid was prepared by dissolving an oxidoreductase in 0.1% CHAPS.

$[\text{Ru}(\text{NH}_3)_6]\text{Cl}_3$ (Dojin Chemical Laboratory "LM722") was used as the electron transporter, and PGGGDH (800 U/mg

glucose dehydrogenation activity) was used as the oxidoreductase.

(1) SWN solution		(2) CHAPS solution		(3) Distilled water	(4) ACES solution	
Concentration	Content	Concentration	Content		Concentration	Content
1.2%	250 μ L	10%	25 μ L	225 μ L	200 mM	500 μ L

In Table 1 and the like, SWN is an abbreviation for Lucentite SWN, CHAPS is an abbreviation for 3-[(3-cholamidopropyl)dimethylammonio]propanesulfonic acid, and ACES is an abbreviation for N-(2-acetamido)-2-aminoethanesulfonic acid. Coop Chemical "3150" was used as the SWN, Dojin Chemical Laboratory "KC062" as the CHAPS and Dojin chemical Laboratory "ED067" as the ACES. The ACES solution was adjusted to a pH of 7.5.

Water-absorbing layer 51 was formed to a thickness of 2 μ m by first applying 0.1 μ L of a coating material comprising an absorbent polymer to the target site of cover 5 and then blow drying it (30°C, 10% Rh). 7 parts by weight of absorbent polymer (Sumitomo "Aquacork") dissolved in 100 parts by weight of methanol was used as the coating material.

(Measuring response current)

Using a biosensor constructed as described above, response current was measured as a time course for three types of blood (Hct 20%, Hct 42% and Hct 69%) with

different hematocrit (Hct) values and a glucose concentration of 447 mg/dL. Response current was measured 5 times for each Hct type of blood. The amount of blood supplied to internal channel 60 of biosensor 1 here was 0.5 μ L, and the voltage applied between working electrode 31 and counter electrode 32 was 500 mV. The results are shown in Figure 11.

The effect of Hct was investigated based on response current values 5 seconds after the supply of blood. The results are shown in Figure 13. In Figure 13 Hct (%) is shown on the horizontal axis and percent bias from response current at Hct 42% on the vertical axis. In Figure 13, percent bias is shown as the average of 5 measurements.

Comparative Example 1

In this comparative example a biosensor was used having the absorbent layer omitted from the biosensor of Example 1, and a time course of response current values was measured for three types of blood of differing Hct as in Example 1. Response current was measured 5 times for each Hct type of blood. The results are shown in Figure 12. Also as in Example 1, the effect of Hct was investigated based on response current values 5 seconds after beginning blood supply. The results are shown in Figure 14.

Discussion of Experimental Results

As can be seen from Figures 11 and 12, the response current values tended to converge more quickly with the biosensor of Example 1 than with the biosensor of Comparative Example 1 when blood of different Hct was measured. First, as can be seen from looking at the 5-second values for response current for example, the difference in response currents between Hct 20% and Hct 69% was smaller with the biosensor of Example 1 than with the biosensor of Comparative Example 1, and second, the response currents for each sample had become uniform after about 8 seconds with the biosensor of Example 1. With the biosensor of Comparative Example 1, by contrast, it took about 15 seconds for the response currents to become uniform.

These results mean that the biosensor of Example 1 is capable of performing proper concentration measurements in less time than the biosensor of Comparative Example 1. Moreover, because in the biosensors of Example 1 and Comparative Example 1 the internal channel 60 of capillary 6 has a small volume of 0.5 μ L, it appears that the biosensor of Example 1 can precisely measure tiny quantities of blood.

Although reproducibility is less with the biosensor of Example 1 than with the biosensor of Comparative Example 1, this is probably attributable to variation in the formation of water-absorbing layer 51 because the coating material was applied by hand to form water-absorbing layer 51. Consequently, reproducibility could

probably be improved if water-absorbing layer 51 could be formed uniformly.

As can be seen from Figures 13 and 14, the biosensor of Example 1 had a bias of +5% in response current 5 seconds after initiation of blood supply within the range of Hct 20-69%, while the biosensor of Comparative Example 1 had a bias of $\pm 20\%$. This means that in the biosensor of Example 1 Hct affects response current less than it does in the biosensor of Comparative Example 1. These results show that the effect of blood Hct is reduced by the inclusion of a water-absorbing layer 51 such as that of the biosensor in Example 1.

As explained above, with the measuring instrument of the present invention it is possible to precisely measure concentration in a small quantity of sample liquid while controlling the effect of the solid components in the sample liquid and keeping the measurement time short.